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Search Results -

Term	Documents
EIGENVECTOR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	725
EIGENVECTORS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	869
EIGENVALUE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1210
EIGENVALUES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1244
(54 AND (EIGENVECTOR OR EIGENVALUE)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l54 and (eigenvector or eigenvalue)	1	L55
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l53 and (matrix)	4	L54
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l52 and (covariance)	5	L53
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l34 and (image with contrast)	85	L52
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l49 and (rank$4)	0	L51
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l49 and (eigenvector or eigenvalue)	0	L50
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l47 and (covariance)	1	L48
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l46 and (image with contrast)	50	L47
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l45 and (image)	223	L46
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (spectral with band$4)	488	L45
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l39 and ((magnetic adj resonance) or MRI or NMR)	1	L44
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l40 and ((magnetic adj resonance) or MRI or NMR)	0	L43
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l41 and ((magnetic adj resonance) or MRI or NMR)	0	L42
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l40 and (image with contrast)	1	L41
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l38 and (matrix)	19	L39
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l36 and (orthogonal$2 or perpendicular$2)	28	L37
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (rank$4)	1690	L34
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l31 and (direction with scan$5)	0	L32
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l28 and ((combin$6 or fus$4) with image)	22	L31
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l29 and((plurality or multiple or multi) with slice)	5	L30
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l28 and (gradient)	21	L29
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l26 and (correlat$6)	32	L27
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l25 and (direction)	61	L26
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l24 and ((magnetic adj resonance) or MRI or NMR)	73	L25
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	(in-plane with resolution)	218	L24
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l22 and (register$4)	6	L23
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l21 and ((magnetic adj resonance) or MRI or NMR)	26	L22
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l20 and (resolution or resolv$6)	43	L21
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l19 and (correlat$6)	62	L20
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l17 and (image)	333	L18
	l16 and (((plurality or multiple or multi)		

USPT,PGPB,JPAB,EPAB,DWPI,TDBD	with slice) with (second or additional or another or first))	699	L17
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	((plurality or multiple or multi) with slice)	4541	L16
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l13 and (scan$5 with direction)	5	L15
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l12 and (resolution or resolv$9)	52	L13
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (correlat$6 with gradient)	113	L12
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l9 and (resolution or resolv$6)	2	L11
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l9 and (min$6)	0	L10
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l8 and (max$6)	3	L9
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (hill-climb$4 or "hill climb$4")	2	L6
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l4 and (hill-climb$4 or "hill climb$4")	0	L5
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l3 and ((((two or another or second) with scan$4) with direction) with (orthogonal$2 or perpendicular$3))	41	L4
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l2 and (((two or another or second) with scan$4) with direction)	204	L3
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and ((two or another or second) with scan$4)	2571	L2
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	((magnetic adj resonance) or MRI or NMR)	121812	L1

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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l22 and (register$4)	6	L23
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l21 and ((magnetic adj resonance) or MRI or NMR)	26	L22
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l20 and (resolution or resolv$6)	43	L21
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l19 and (correlat$6)	62	L20
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l18 and (combin$6)	211	L19
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l17 and (image)	333	L18
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l16 and (((plurality or multiple or multi) with slice) with (second or additional or another or first))	699	L17
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	((plurality or multiple or multi) with slice)	4541	L16
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l13 and (scan$5 with direction)	5	L15
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (correlat$6 with gradient)	113	L12
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l9 and (min$6)	0	L10
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l8 and (max$6)	3	L9
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l2 and (hill-climb$4 or "hill climb$4")	0	L7
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and (hill-climb$4 or "hill climb$4")	2	L6
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l4 and (hill-climb$4 or "hill climb$4")	0	L5
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USPT,PGPB,JPAB,EPAB,DWPI,TDBD	l1 and ((two or another or second) with scan$4)	2571	L2
USPT,PGPB,JPAB,EPAB,DWPI,TDBD	((magnetic adj resonance) or MRI or NMR)	121812	L1

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Search Results - Record(s) 1 through 6 of 6 returned.

1. Document ID: US 6306092 B1

L8: Entry 1 of 6 File: USPT Oct 23, 2001

US-PAT-NO: 6306092
DOCUMENT-IDENTIFIER: US 6306092 B1

TITLE: Method and apparatus for calibrating rotational offsets in ultrasound transducer scans

DATE-ISSUED: October 23, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Yamrom; Boris	Schenectady	NY		
Hatfield; William Thomas	Schenectady	NY		
Piel, Jr.; Joseph Edward	Scotia	NY		
Avila; Ricardo Scott	Clifton Park	NY		

US-CL-CURRENT: 600/447; 128/916, 29/25.35

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

2. Document ID: US 6259943 B1

L8: Entry 2 of 6 File: USPT Jul 10, 2001

US-PAT-NO: 6259943
DOCUMENT-IDENTIFIER: US 6259943 B1

TITLE: Frameless to frame-based registration system

DATE-ISSUED: July 10, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Cosman; Eric R.	Belmont	MA		
Cundari; Michael A.	Hingham	MA		
Ledoux; Robert J.	Bedford	MA		
Daniels; Robert A.	Haverhill	MA		
Mauge; Christophe P.	Somerville	MA		
Labuz; Jeffrey	Brookline	MA		

US-CL-CURRENT: 600/429; 600/417, 606/130

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

3. Document ID: US 6066084 A

L8: Entry 3 of 6 File: USPT May 23, 2000

US-PAT-NO: 6066084
DOCUMENT-IDENTIFIER: US 6066084 A

TITLE: Method and apparatus for focused neuromagnetic stimulation and detection

DATE-ISSUED: May 23, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Edrich; Jochen	Neu-Ulm			DEX
Zhang; Tongsheng	Neu-Ulm			DEX

US-CL-CURRENT: 600/13

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw. Desc | Image

4. Document ID: US 5768405 A

L8: Entry 4 of 6 File: USPT Jun 16, 1998

US-PAT-NO: 5768405
DOCUMENT-IDENTIFIER: US 5768405 A

TITLE: Digital image processing method for local determination of the center and the width of objects in the form of contrasting bands on a background

DATE-ISSUED: June 16, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Makram-Ebeid; Sherif	Dampierre			FRX

US-CL-CURRENT: 382/128; 382/264, 382/265

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw. Desc | Image

5. Document ID: US 5703960 A

L8: Entry 5 of 6 File: USPT Dec 30, 1997

US-PAT-NO: 5703960
DOCUMENT-IDENTIFIER: US 5703960 A

TITLE: Lumber defect scanning including multi-dimensional pattern recognition

DATE-ISSUED: December 30, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Soest; Jon F.	Seattle	WA		

US-CL-CURRENT: 382/141; 250/559.18, 250/559.45, 348/91, 348/92, 382/108, 382/110

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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6. Document ID: TW 245805 A

L8: Entry 6 of 6   File: DWPI   Apr 21, 1995

DERWENT-ACC-NO: 1995-205555
DERWENT-WEEK: 199527

TITLE: Conjugate radio-frequency coil system for high resolution local brain MRI - uses differently orientated single coils to receive electrical signals induced from RF magnetic fields and obtain local brain images

INVENTOR: GUAN, W

PRIORITY-DATA: 1994TW-0106663 (July 21, 1994)

PATENT-FAMILY:

PUB-NO	PUB-DATE	LANGUAGE	PAGES	MAIN-IPC
TW 245805 A	April 21, 1995		017	H01F005/00

INT-CL (IPC): H01F 5/00

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Clip Img | Image

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Term	Documents
CORRELAT$6	0
CORRELAT.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	48
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CORRELATABLE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1048
CORRELATABLY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	5
CORRELATAIVE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
CORRELATAR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
CORRELATATE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
CORRELATD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	5
CORRELATDD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
(L4 AND (CORRELAT$6)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	6

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L8: Entry 1 of 6 File: USPT Oct 23, 2001

DOCUMENT-IDENTIFIER: US 6306092 B1
TITLE: Method and apparatus for calibrating rotational offsets in ultrasound transducer scans

ABPL:
The axis of rotational transducer array scans, because of imperfect transducer array assembly, may have two orthogonal offsets relative to the geometric center of the transducer array. Without knowledge of these offsets, it is not possible to convert rotational transducer scan data into a rectilinear (Euclidean) coordinate system, as is necessary for three-dimensional processing. Using spatial coherency between appropriate scan lines in different rotational transducer scans, the horizontal and vertical rotational offsets are calculated. These offsets are then utilized in converting the data to a rectilinear coordinate system for three-dimensional processing.

BSPR:
Three-dimensional visualization has gained popularity in medical applications since the introduction of computer tomography (CT) in the field several decades ago. For example, three-dimensional visualization is also used in magnetic resonance (MR) imaging. Using three-dimensional data sets in ultrasound imaging is not as popular due to two major obstacles: first, data in most cases are acquired by free-hand B-mode scans that do not provide sufficiently accurate information to enable precise positioning of individual two-dimensional scans (slices) into a common three-dimensional coordinate space; second, the ultrasound data are inherently more noisy than CT and MR data sets, and therefore traditional surface visualization techniques do not produce good results. The last decade has brought many advances in technology, in both hardware and software, that allow for real-time three-dimensional data set visualization using so-called volume rendering that goes directly from a three-dimensional data set into a two-dimensional image display, bypassing the creation of surfaces. One volume rendering technique is known as maximum intensity projection (MIP). The MIP technique involves projection of three-dimensional data intensity values onto a two-dimensional image plane by assigning to each image pixel the maximum intensity value in the three-dimensional data volume that belongs to the line of sight that goes from the eye point through this pixel and into the volume. This method, in combination with animation, can produce true three-dimensional impressions on the monitor. A more computationally demanding technique is known as compositing. This technique involves modeling of a physical phenomenon of light propagation in semi-translucent/semi-opaque media that is recreated from a three-dimensional data set with the addition of specially designed transfer functions.

BSPR:
Some medical applications involve acquiring three-dimensional volume data by a transducer that rotates around an axis orthogonal to the transducer array. The volume "swept" by these two-dimensional B-mode scans represents a cylinder. Since the two-dimensional scans do not lie parallel to each other, it is difficult to visualize three-dimensional object structures from individual scans alone and a volume visualization technique would be desirable. Contemporary software and hardware are efficient in volume rendering techniques, but require that the data be represented as a rectilinear three-dimensional data array. Therefore, conversion from a cylindrical coordinate system to a rectilinear coordinate system is required. Although this conversion is not difficult to compute, an important practical complication to the conversion process is that there is always some offset of the axis of rotation relative to the sensor array middle point. Need exists

for a method of calculating this offset based on one three-dimensional volume scan. The offset information computed can be used in an algorithm to convert from a cylindrical coordinate system to a rectilinear one and also can be used in the transducer manufacturing process to position a transducer array exactly at the rotational axis.

DRPR:
FIG. 9 is a schematic illustration of two overlapping coordinate grids in a plane orthogonal to the rotational transducer scans: one induced by rotational scans and the other induced by parallel scans.

DRPR:
FIGS. 17 and 18 are schematic representations of two consecutive rotational transducer scans.

DRPR:
FIG. 19 is a graph of the correlation function between corresponding vertical scan lines in two consecutive rotational transducer scans.

DRPR:
FIGS. 22 and 23 are schematic representations of two rotational transducer scans corresponding to slices PQ and P'Q', respectively, in FIG. 21.

DRPR:
FIG. 24 is a graph of the correlation function between vertical scan lines in two conjugate slices (PQ and P'Q').

DEPR:
Rotational transducer scans are generated by a one-dimensional transducer array mounted on a rotating plate 3 (FIG. 1) having an axis of rotation perpendicular to the transducer array. The invention addresses the situation wherein the center of the transducer array is offset from the axis of rotation of the rotating plate. Each scan represents a two-dimensional data array 36, shown in FIG. 4, which is M elements wide and N elements high. The totality of scans for all rotational angles of the transducer sweeps a cylindrical volume 38, shown in FIG. 3. Various post-processing techniques usually deal with rectilinear structured grid sets. FIGS. 5 and 6, respectively, represent a three-dimensional rectilinear volume data set 40 and one of its two-dimensional horizontal slices 42. The slices are M elements wide and M elements deep. The volume is N elements high. The total amount of elements in the volume data set 40 is M.times.M.times.N . When converting from rotational slices to rectilinear coordinates, sampling over volume M.times.M.times.N need not be performed with the same sampling steps along rectilinear axes. However, without loss of generality, the same number of elements along horizontal rectilinear axes will be used as the number of elements in the rotational transducer.

DEPR:
In accordance with a preferred embodiment of the invention, the procedure for computing offset_x is as follows. In FIG. 15, the lines PQ and P'Q' represent top views of two consecutive rotational transducer scans. FIG. 16 shows the center of rotation O, projection S of center of rotation O on scan PQ and projection S' of center of rotation O on scan P'Q'. FIGS. 15 and 16 also show the intersection T of these two scans. For two consecutive scans, distances ST and S'T, shown in FIG. 16, will be negligibly small (they are equal as follows from geometry) . Also, distance PS is equal to distance P'S' from the definition of O as the rotational center. Therefore distance PT can be assumed to be approximately equal to distance P'T. Since T is a point shared by two rotational transducer scans, a scan line passes through T perpendicular to the plane of the illustration and is shared by those two rotational transducer scans. Therefore, a comparison of the data in two scan lines S in two consecutive scans (vertical scan lines S in both scans shown in FIGS. 17 and 18), will show one value S, 1.ltoreq.S.ltoreq.M, for which these two data arrays will be most correlated (the presence of noise prevents these data arrays from being equal, even if they represent scan data for one and the same line in physical space). Here any correlation function can be used. For two data arrays X=(x.sub.1, . . . , x.sub.n), Y=(y.sub.1, . . . , y.sub.n), the formula ##EQU1##

DEPR:
The graph of FIG. 19 shows the generic view of this correlation coefficient as a function of scan line number S. In this instance, X is the vertical scan line number S in the scan k and Y is the vertical scan line number S in the scan (k+1), k=1 to 178. For each number k, S(k,k+1) is computed as a scan line number which maximizes the correlation coefficient. For multiple maxima, there are several choices: a) take the minimum of those S maximizing correlation; b) take the maximum of S; and c) take average of S. For example, if the total number of vertical scan lines in each scan were 11, then using method a) for a list of correlation coefficients {0.2 0.3 0.2 0.2 0.4 0.5 0.8 0.8 0.7 0.6 0.4} for scan lines 1 to 11 in scans 5 and 6, S(5,6)=7 is first computed. Then the average of all such S(k,k+1) is computed: ##EQU2##

DEPR:
To compute offset_y, a preferred embodiment utilizes the coherency between far-spaced scans. The doubly cross-hatched area in FIG. 14 represents area swept twice: once by the leftmost part PO" and once by the rightmost part O"Q of the transducer array. FIG. 20 shows that the sought offset is equal to the distance O'O. For all the points T of the doubly cross-hatched area of FIG. 14 lying at the same horizontal level as 0', two slices PQ and P'Q' are found intersecting at T (see FIG. 21). If slice PQ has scan number k, then slice P'Q' will have scan number (180-k), since .angle.QTO'=.angle.O'TP'. Knowing that P'O"=PO=&Scirc ;, then P'T=&Scirc ;+O"T and PT=&Scirc ;-OT. The fact that OT=TO" leads to the following algorithm for computing offset_y. Defining slices number k and number (180-k) as conjugate slices, then for all conjugate pairs of slices, a correlation coefficient can be computed for the vertical scan line &Scirc ;-OT in slice k shown schematically in FIG. 22, and the vertical scan line &Scirc ;+OT in slice (180-k) shown schematically in FIG. 23. A graph for this correlation coefficient as a function of OT for all possible values of OT can be plotted as shown in FIG. 24. Since the diagram of FIG. 14 represents one of four different combinations of signs of offset_x and offset_y, the value of offset OT (T in FIGS. 21 and 22) should be allowed to be both positive and negative, with the condition that it has an opposite sign in the conjugate slice. As previously with offset_x, the offset is selected that maximizes the correlation coefficient. Now, if offset OT corresponds to the maximum correlation coefficient, then for the pair of conjugate slices k and (180-k)

CLPR:
16. The method as recited in claim 15, wherein the step of employing the acquired scan data to calculate offset data comprises employing the acquired scan data to calculate a first offset along a first coordinate direction and a second offset along a second coordinate direction perpendicular to said first coordinate direction.

CLPR:
18. The method as recited in claim 17, wherein the step of employing the acquired scan data to calculate a second offset comprises the steps of:

CLPV:
means for calculating correlation coefficients for respective scan lines of each pair of rotational scans of said transducer array having scan numbers k and (k+1), where k varies from 1 to K and K is the total number of scans minus 1;

CLPV:
means for determining a scan line number S(k,k+1) corresponding to a maximum correlation coefficient for each scan number k; and

CLPV:
means for calculating correlation coefficients for a vertical scan line &Scirc ;-OT in scan k and a vertical scan line &Scirc ;+OT in a scan conjugate to scan k for each pair of conjugate rotational transducer array scans, where line segment OT is defined by a point T where said pair of conjugate scans intersect and point O where a line OO' perpendicular to scan k and intersecting said center of rotation O' intersects scan k; and

CLPV:
means for determining a value for OT corresponding to a maximum correlation

coefficient for each said pair of conjugate scans;

CLPV:
wherein said second offset is calculated as a function of said value for OT corresponding to a maximum correlation coefficient.

CLPV:
calculating correlation coefficients for respective scan lines of each pair of rotational transducer scans having scan numbers k and (k+1), where k varies from 1 to K and K is the total number of scans minus 1;

CLPV:
determining a scan line number S(k,k+1) corresponding to a maximum correlation coefficient for each scan number k;

CLPV:
calculating correlation coefficients for a vertical scan line &Scirc ;-OT in scan k and a vertical scan line &Scirc ;+OT in a scan conjugate to scan k for each pair of conjugate rotational transducer array scans, where line segment OT is defined by a point T where said pair of conjugate scans intersect and point O where a line OO' perpendicular to scan k and intersecting said center of rotation O' intersects scan k;

CLPV:
determining a value for OT corresponding to a maximum correlation coefficient for each said pair of conjugate scans; and

CLPV:
calculating a second offset as a function of said value for OT corresponding to a maximum correlation coefficient.

CLPV:
calculating correlation coefficients for respective scan lines of each pair of rotational scans of said transducer array having scan numbers k and (k+1), where k varies from 1 to K and K is the total number of scans minus 1;

CLPV:
determining a scan line number S(k,k+1) corresponding to a maximum correlation coefficient for each scan number k;

CLPV:
calculating correlation coefficients for a vertical scan line &Scirc ;-OT in scan k and vertical scan line &Scirc ;+OT in the scan conjugate to scan k for each pair of conjugate rotational transducer array scans, where line segment OT is defined by a point T where said pair of conjugate scans intersect and point O where a line OO' perpendicular to scan k and intersecting said center of rotation O' intersects scan k;

CLPV:
determining a value for OT corresponding to a maximum correlation coefficient for each said pair of conjugate scans; and

CLPV:
calculating a second offset as a function of said value for OT corresponding to a maximum correlation coefficient.

CLPV:
calculating correlation coefficients for respective scan lines of each pair of rotational scans of said transducer array having scan numbers k and (k+1), where k varies from 1 to K and K is the total number of scans minus 1;

CLPV:
determining a scan line number S(k,k+1) corresponding to a maximum correlation coefficient for each scan number k;

CLPV:
calculating correlation coefficients for a vertical scan line &Scirc ;-OT in scan k and a vertical scan line &Scirc ;+OT in a scan conjugate to scan k for each pair of conjugate rotational transducer array scans, where line segment

OT is defined by a point T where said pair of conjugate scans intersect and point O where a line OO' perpendicular to scan k and intersecting said center of rotation O' intersects scan k;

CLPV:
determining a value for OT corresponding to a maximum correlation coefficient for each said pair of conjugate scans; and

CLPV:
calculating a second offset as a function of said value for OT corresponding to a maximum correlation coefficient.

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☐ 1. Document ID: US 6317619 B1

L23: Entry 1 of 6 File: USPT Nov 13, 2001

US-PAT-NO: 6317619
DOCUMENT-IDENTIFIER: US 6317619 B1

TITLE: Apparatus, methods, and devices for magnetic resonance imaging controlled by the position of a moveable RF coil

DATE-ISSUED: November 13, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Boernert; Peter	Hamburg			DEX
Schaeffter; Tobias Richard	Hamburg			DEX
Weiss; Steffen	Hamburg			DEX

US-CL-CURRENT: 600/410; 324/307, 324/309, 324/318, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KWIC | Draw. Desc | Image

☐ 2. Document ID: US 5787889 A

L23: Entry 2 of 6 File: USPT Aug 4, 1998

US-PAT-NO: 5787889
DOCUMENT-IDENTIFIER: US 5787889 A

TITLE: Ultrasound imaging with real time 3D image reconstruction and visualization

DATE-ISSUED: August 4, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Edwards; Warren	Seattle	WA		
Deforge; Christian	Seattle	WA		
Kim; Yongmin	Seattle	WA		

US-CL-CURRENT: 600/443; 128/916

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KWIC | Draw. Desc | Image

☐ 3. Document ID: US 5488952 A

L23: Entry 3 of 6 File: USPT Feb 6, 1996

US-PAT-NO: 5488952
DOCUMENT-IDENTIFIER: US 5488952 A

TITLE: Stereoscopically display three dimensional ultrasound imaging

DATE-ISSUED: February 6, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Schoolman; Arnold	Kansas City	MO		

US-CL-CURRENT: 600/443; 128/916

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KWIC | Draw Desc | Image

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☐ 4. Document ID: US 5239591 A

L23: Entry 4 of 6 File: USPT Aug 24, 1993

US-PAT-NO: 5239591
DOCUMENT-IDENTIFIER: US 5239591 A

TITLE: Contour extraction in multi-phase, multi-slice cardiac MRI studies by propagation of seed contours between images

DATE-ISSUED: August 24, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ranganath; Surendra	Peekskill	NY		

US-CL-CURRENT: 382/128; 345/442, 382/199, 382/302, 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

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☐ 5. Document ID: US 5086275 A

L23: Entry 5 of 6 File: USPT Feb 4, 1992

US-PAT-NO: 5086275
DOCUMENT-IDENTIFIER: US 5086275 A

TITLE: Time domain filtering for NMR phased array imaging

DATE-ISSUED: February 4, 1992

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Roemer; Peter B.	Schenectady	NY		

US-CL-CURRENT: 324/309; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

☐ 6. Document ID: US 4945478 A

L23: Entry 6 of 6 File: USPT Jul 31, 1990

US-PAT-NO: 4945478
DOCUMENT-IDENTIFIER: US 4945478 A

TITLE: Noninvasive medical imaging system and method for the identification and 3-D display of atherosclerosis and the like

DATE-ISSUED: July 31, 1990

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Merickel; Michael B.	Charlottesville	VA		
Carman; Charles S.	Charlottesville	VA		
Brookeman; James R.	Charlottesville	VA		
Mugler, III; John P.	Charlottesville	VA		
Ayers; Carlos R.	Charlottesville	VA		

US-CL-CURRENT: 382/131; 382/180

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw. Desc | Image

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L23: Entry 5 of 6 File: USPT Feb 4, 1992

DOCUMENT-IDENTIFIER: US 5086275 A
TITLE: Time domain filtering for NMR phased array imaging

ABPL:
A method and apparatus for combining NMR response data of a sample from a plurality of closely spaced RF receiver coils of an NMR phased array in the time domain to form a composite NMR image wherein each of the RF receiver coils receives a different respective one of a plurality of NMR response signals, each of which is evoked from a portion of the sample within a field of a respective one of the receiver coils. The response signals are conditioned to develop a plurality of data point signals corresponding to the magnitude of each of the respective response signals from each of the receiver coils at successive time intervals. The data point signals are convolved by a time domain representation of a field map of the respective one of the receiver coils generating the corresponding one of the response signals. The convolved signals are combined on a time domain point-by-point basis to produce a time domain representation of the composite NMR image of the sample.

BSPR:
The present invention relates to nuclear magnetic resonance (NMR) imaging and, more particularly, to methods and apparatus for combining the simultaneously received data from a plurality of radio-frequency (RF) coils of an NMR phased array in the time, rather than image, domain to produce a composite image having high signal-to-noise ratio (SNR) throughout the image.

BSPR:
The term "NMR phased array" refers to apparatus, such as shown in Roemer et al. U.S. Pat. No. 4,871,969 (the disclosure of which is incorporated herein by reference), wherein a plurality of closely-spaced RF coils is employed for simultaneously receiving different NMR response signals from associated portions of a sample (such as a patient in medical imaging) and combining the separate data from each coil to produce a single composite NMR image of the sample. By overlapping adjacent coils and connecting each coil to the input of an associated low-input-impedance preamplifier channel, the high SNR of a single surface coil can be maintained over fields-of-view (FOV) characteristic of remote coils.

BSPR:
Currently, composite images for NMR phased arrays are reconstructed in the image domain by combining the individual image contributions on a weighted, point-by-point basis after first acquiring the complete NMR images for each separate coil. The reason for acquiring the separate images first is that the optimum set of weights needed to maximize SNR when combining the separate signals to produce the composite image is a function of position, and so varies from point to point. While the phase shifters and transformers of the setup shown in FIG. 6 of Roemer et al. U.S. Pat. No. 4,825,162 can be adjusted to provide a composite image in the time domain having a high SNR at any particular point, different weighting must be applied for each point in order to obtain good sensitivity over the whole image. Thus, the conventional approach is to first separately acquire the different NMR image from each coil before combining the different individual images, on a point-by-point basis, to form the composite image.

BSPR:
NMR phased array imaging as described in the '162 patent, therefore, has the drawbacks of requiring large amounts of memory to store the separate coil

images before reconstruction and of necessitating long time delays between acquisition of the last data point and onset of the first display of the reconstructed image.

BSPR:
It is desirable in NMR phased array imaging to be able to combine the data from the separate receiver coils as it is acquired on a time domain, rather than image domain, basis without sacrificing SNR resolution. Combining the data as acquired will reduce the total memory requirements of the system since only one combined data set would have to be stored and, because only the combined data set will have to be transformed at the end of scanning, will also reduce the time between end-of-scan and first appearance of the composite image.

BSPR:
Among the several objects of the present invention will be noted the provision of a method and apparatus for forming a composite NMR image with high SNR throughout the image; the provision of a method and apparatus for NMR spectroscopy and NMR imaging using data combination in real time; and the provision of a method and apparatus which overcomes the aforementioned and other disadvantages of the prior art.

BSPR:
In accordance with the invention, a method is provided for combining the simultaneously received different NMR response signals from a plurality of closely-spaced, overlapping RF receiver coils of an NMR phased array in the time domain, to form a composite image that has high SNR throughout the image. A filter scheme is utilized to develop a composite data set in the time domain, wherein each time point of the composite data is formed on the basis of contributions from previous data points and future data points. The data is passed through filter arrangements having one-, two- and three-dimensional filters before the signals are summed together. Each filter dimension corresponds to filtering in one of the time dimensions of k-space, i.e., the readout direction; the phase encode direction; and, in the case of three-dimensional imaging, the second phase encode direction. Filter coefficients are chosen to combine the data in a way that is simultaneously optimal for providing a high SNR at multiple points of the composite image. With more terms added to the filter, the SNR can be optimized over the entire image.

DRPR:
FIG. 1 (prior art) is a schematic view of an arrangement employed in the conventional method to combine the signals from the overlapping coils of an NMR phased array using image domain data processing techniques.

DRPR:
FIG. 4 is a filter function derived for the coil of FIG. 2 after Fourier transformation into image space of the truncated representation of FIG. 3.

DRPR:
FIG. 5 is a schematic view of an arrangement employed in a method of combining the coil signals of an NMR phased array in the time domain using filters in accordance with the present invention.

DRPR:
FIG. 6 is a pictorial representation of truncated convolution in the arrangement of FIG. 5 for a two-dimensional single slice image.

DRPR:
FIGS. 10A, 10B, 10C and 10D are views of exemplary reconstructions in the image space (FIG. 10A) and time domain space (FIGS. 10B-10D), respectively, showing the effect of filtering in accordance with the method of the invention.

DEPR:
FIGS. 1 and 5 show an NMR phased array 10, such as described in Roemer et al. U.S. Pat. No. 4,825,162, of a plurality of radio-frequency (RF) receiver coils 12 (coils 1 through N.sub.c) defining an imaging volume for the NMR imaging of a sample, such as for the NMR medical diagnostic imaging of a human spine. The

separate surface coils 12 are identically configured and are arranged in closely-spaced relationship with overlapping fields-of-view (FOV), but with substantially no interaction between adjacent coils. The coils 12 are adapted as part of the NMR imaging process to simultaneously receive a different one of a plurality of NMR response signals each evoked from an associated portion of the sample enclosed in the imaging volume. As shown, each coil 12 has its own processing channel 14 including receiver circuitry 15 and an analog-to-digital converter 16. FIG. 1 is a schematic representation of the conventional data processing set-up for constructing a different NMR image for each channel 14 of a sample portion from the NMR response signals received by the associated coil 12 for that channel 14, and for subsequently combining the plurality of different images thus constructed, on a point-by-point basis, in the image domain, to produce a single final NMR image of all sample portions from which an NMR signal was received by any of the coils 12. FIG. 5 is a schematic representation of the corresponding set-up for performing the image reconstruction in the time domain utilizing the principles of the present invention.

DEPR:
As described in Roemer et al. U.S. Pat. No. 4,871,969, the optimal combination or weighting of signals from the individual coils 1-N.sub.c in the array 10 to achieve a high signal-to-noise ratio (SNR) is dependent on the location (x, y, z) of a particular volume element (voxel). This is because the signal of each RF receiving coil C.sub.i is sensitive to nuclear spins in proportion to the field B.sub.i created by the coil, whereas the noise is "white noise" uniformly distributed over the image. Hence, the resultant SNR is a function of position.

DEPR:
Assume that I.sub.1 (x,y) is the complex image obtained by reconstructing the data received from coil C.sub.i, and B.sub.i (x,y) is the RF magnetic field produced by coil C.sub.i. The real part of B is the x component (in magnet coordinates as opposed to the screen coordinates of the image) of the transverse RF magnetic field and the imaginary part of B is the y component of the field. If noise correlations are ignored (which will have little effect on image quality) and all coils 12 have approximately the same noise, the combination of separate images I.sub.i that optimizes the SNR in the composite image is given by ##EQU1## where I(x,y) is the composite image.

DEPR:
The complex image is really the product of the RF receiving coil magnetic field and the spin density S(x,y) given by

DEPR:
The magnitude of I(x,y) in equation (1) can be expressed as the sum of the products of the magnitudes of the image and the magnetic field maps: ##EQU2##

DEPR:
Equation (3) gives a basic form usable in image reconstruction methods. Combining images using equation (3) is particularly convenient because the phase shifts of the individual receivers do not have to be known, and the image reconstruction programs do not have to carry the complex data.

DEPR:
FIG. 1 shows schematically the conventional process employed for combining the data in the image domain. The NMR signal from each coil 1-N.sub.c is sent through its associated channel 14 for processing by its own receiver 15, digitized by its own analog-to-digital (A/D) converter 16 and stored in digital form in its own assigned memory 18. After acquisition is complete, the data from each coil channel is separately subjected to transformation by processing means 19 and then combined point-by-point into a single composite image at summation means 20 in accordance with equation (3).

DEPR:
To derive the time domain filtering method of the present invention (FIG. 5), it was recognized that the combined image obtained using the image domain method is simply the Fourier transform of the original time dependent data. In accordance with equation (1), the optimal combination of images I(x,y) (i.e., that giving high SNR over the whole image) is obtained by multiplying each

separate coil image I.sub.i (x,y) by its corresponding RF coil magnetic field map profile B.sub.i (x,y) at 21, before summing the results at 20 (see FIG. 1). From linear system theory, however, it is known that convolution in the time domain is equivalent to multiplication in the spatial domain. Thus, for a single slice of multi-slice data the time domain representation of the composite image can be given by the two-dimensional convolution integral ##EQU3## where N.sub.c is the number of coils, v.sub.i (t.sub.r,t.sub..PHI.) is the time dependent NMR voltage signal measured on coil i, b.sub.i (.tau..sub.r,.tau..sub..PHI.) is the inverse Fourier transform of coil i's RF field map, A(t.sub.r,t.sub..PHI.) is the inverse Fourier transform of the composite data set and t.sub.r is the readout time for each phase encode time t.sub..PHI..

DEPR:
For a finite set of discrete samples, the inverse Fourier transform A(j,k) of the composite image is ##EQU4## where v.sub.i (j,k) is a matrix of NMR voltages measured on coil i and b.sub.i (j,k) is the discrete Fourier transform of the field map from coil i. The first and second arguments are the sample indices in the readout and phase encode directions, respectively, and N.sub.r and N.sub..PHI. are the number of samples in the readout and phase encode directions, respectively.

DEPR:
At first glance, equation (5) which combines the data in the time domain does not appear to offer any computational advantages over equation (3) which combines the data in the image domain. According to equation (5), the total number of operations required to obtain a single k-space data point of the composite image is proportional to the number of pixels N in the image, where N=N.sub..PHI. .times.N.sub.r. Thus, the number of operations required to construct the entire k-space representation of the composite image scales is a factor of N.sup.2. So, for large values of N, combining the data in image space rather than in time space would appear to require far fewer computations. This is because the data from each coil channel is subjected to Fourier transformation at 19 (FIG. 1) before combining as a simple weighted sum at 20, and the number of operations for a Fast Fourier transform (FFT) scales as a factor of N log(N) rather than N.sup.2.

DEPR:
The number of computations required for the convolution can be greatly reduced, however, through the recognition that the RF field map 21 (shown in FIG. 1) are relatively slowly varying quantities across the image and can thus be suitably represented in abbreviated form for time domain processing purposes. It has been observed that the inverse Fourier transform of the field map is concentrated near the origin in the time domain (k-space) and thus the b.sub.i (l,m) terms in equation (5) can be truncated to a kernel containing relatively few terms.

DEPR:
By way of example, FIG. 2 shows the magnitude of a calculated sensitivity profile of a typical surface coil. The calculation is for a 40 cm FOV with a 12 cm square loop RF receiving coil located in a plane perpendicular to the image. The main magnetic field is horizontal. The magnitude of its corresponding inverse Fourier transform is shown in FIG. 3, which is a contour plot of the k-space representative of the field map of FIG. 2. The constant contours are designated by arbitrary numerical units scaled to a maximum of 1.0, with the maximum being at the origin. Only the center 31.times.31 pixels of the magnitude are shown. Although the sensitivity profile (FIG. 2) occupies about 24% of the image FOV, the k-space representation (FIG. 3) has significant magnitude contribution only in the center few pixels. FIG. 4 shows the construction of a filter function profile corresponding to the sensitivity profile of FIG. 2, after truncating the filter coefficient of FIG. 3 by setting the magnitudes of the k space representation of FIG. 3 to zero outside the central 9.times.9 pixel matrix of points, placing a Hamming window (see, R. W. Hamming, Digital Filters [Hall] pp. 102-105) around the data to avoid ringing, and then Fourier transforming the result into image space. A visual comparison of the derived filter function profile of FIG. 4 with the original profile of FIG. 2 shows little qualitative difference, except near the coil wires themselves. This indicates that a kernel of 9.times.9 pixels is sufficient to give a good reconstruction. The error near the wires (located at

the intersection of the side lobes and the central region of sensitivity) occurs because the RF magnetic field varies rapidly there and thus contains high spatial frequencies. Away from the coil, the RF field varies slowly and the 9.times.9 filter kernel matches the profile well.

DEPR:
FIG. 5 shows a system and process in accordance with the invention for combining the separate coil data from an NMR array to obtain a composite image with good SNR resolution in the time domain, using the above filtering technique. The front end of each coil channel 14' has its own receiver 15 and A/D circuitry 16 for receiving and digitizing the separately received signal, the same as for the corresponding channels 14 of the image domain processing set-up of FIG. 1. But instead of storing the separate NMR images of each channel in a separate memory location 18 and Fourier transforming at 19 prior to summing, as done in the system of FIG. 1, the arrangement of FIG. 5, in accordance with the invention, filters the data with a field map filter 23 as it is acquired, and then sums the filtered data at summation means 20' prior to storing a pretransformation combined image in a memory 24. A single Fourier transformation is then undertaken by fast Fourier transformation means 25 to give the final composite image. The filters 23 provide the weighting necessary for summing the separate contributions from the channels 14' to give a good SNR resolution in the end image. The filters 23 perform the operations defined by equation (5). In contrast to the image domain data combination method employed by the system of FIG. 1, only one (or, possibly, two to obtain a uniform noise image) Fourier transformation is required at the end of the scanning operation to produce the combined image. Thus the time domain filtering scheme of FIG. 5 avoids the large time delays from end-of-scan to first image appearance inherent in the process employed by the system of FIG. 1. Moreover, the transformation process for the arrangement shown in FIG. 5 is independent of the number of coils utilized. Also, in contrast to the FIG. 1 image domain approach, the time domain method employed by the system of FIG. 5 of the invention has the additional advantage that the data is combined in real time, as it is being collected, thereby reducing the data storage capacity necessary for each coil channel.

DEPR:
Various system architectures are possible for implementation of the time domain image reconstruction method of the invention. If the k-space representation (FIG. 3) of the field map for each coil 12 is reduced to a kernel size of N.sub.cr .times.N.sub.c.PHI. in the readout and phase encode directions (see FIG. 4), the k-space representation of the composite image may be given by ##EQU5## According to equation (6), each data point from each coil 12 contributes to a rectangular subregion of the composite k-space matrix. This relationship is shown pictorially in FIG. 6.

DEPR:
FIG. 6 is a pictorial representation of truncated convolution for a two-dimensional single slice image. The digitized signal 28 from a single coil 12 has a number of data points 29, 30 of magnitude V.sub.n, V.sub.n+1, etc., corresponding to samplings of the analog voltage signal 28 taken at successive time intervals t.sub.n, t.sub.n+1, etc. The data from each coil 12 contributes to a rectangular region in the composite k-space (time domain) matrix. The size of the rectangular region is N.sub.c.PHI. .times.N.sub.cr. As each new data point enters, the subregion moves in the readout direction by one pixel. The signal 29 from time tn contributes to a rectangular subregion 31, and the signal 30 from time t.sub.n+1 contributes to a rectangular subregion 32 shifted in the data memory 24 in the readout direction by one column. For each new phase encode step, the region moves down by one row.

DEPR:
A straightforward hardware embodiment of filter 23 for implementation of equation (6) is depicted in FIG. 7. The arrangement uses discrete components and needs a modest amount of memory to temporarily save the most recent N.sub.c.PHI. .times.N.sub.r data points from each slice, echo and coil. The multiplication element A performs the multiplication in equation (6). The two innermost summations of equation (6) are done by summation element B and the temporary register 36. The outermost sum is performed by summation element 20'. For each coil channel 14', a temporary storage memory 34 is connected to receive the digitized output signal of the associated coil 12. The memory 34

functions to save the data points 29, 30 for the successive time increments tn, tn+1, etc., until a sufficient amount of data is accumulated to complete one i,k point in the composite matrix. Assuming that k-space is covered in a linear fashion, this occurs during the N.sub.c.PHI. 'th phase encode step. As each subsequent data point enters from the left (viz., at the input end), the appropriate N.sub.cr .times.N.sub.c.PHI. data is extracted from filter memory 34 and is multiplied by the corresponding filter coefficients stored in a convolution kernel memory 35, summed by use of a temporary register 36, and then sent to an output port 37 as a weighted input to the composite summation means 20'. To avoid overflow of the input memory 34, a line of readout data must be completed and sent to the output port for each new line that enters. To accommodate multi-slice data, the memory must save the most recent N.sub.c.PHI. .times.N.sub.r data points from each slice and echo.

DEPR:
The architectures shown in FIGS. 7 and 8 do not exploit any special purpose filter chips that might be available. A number of companies make chips that compute one-dimensional convolutions very efficiently, and it may be desirable to utilize such chips to make a compact and efficient filter. FIG. 9 shows one way of breaking the two-dimensional filter 23 (shown in FIG. 5) into a series of one-dimensional filters. If this is done, the data rate into memory can be reduced by a factor of N.sub.c.PHI., the number of convolution filter points in the phase encode direction. For each coil 12, a line of data in the readout direction is passed through a one-dimensional convolution filter 38 a total of N.sub.c.PHI. times. Each pass through the one-dimensional filter utilizes a different set of filter coefficients obtained from the filter memory and corresponds to the multiplication and innermost summation of equation (6). The outermost sum of equation (6), the sum over the coils, is produced by summation means 20'". Summation means 20'" also performs an addition for the read-modify-write operation which is equivalent to the second summation in equation (6). The signals at the outputs 37" of the filters are summed by summation means 20'", and the results are added to the k-space storage memory 24 for the composite images. For each readout line that enters from the left or input side, N.sub.c.PHI. readout lines in the composite stage memory 24 are modified. The same hardware can be used to combine three-dimensional data simply by changing the order in which data is read out of filter memory, and the order in which the results are added to the composite memory.

DEPR:
One proposed digital signal processing chip employs 32 multipliers that operate in parallel and can convolve a 16 bit-by-16 point complex kernel with a 16 bit data path at a rate of 800 ns per complex I/Q pair. For multi-slice data requiring a 16 point-by-16 point kernel and acquired at a 100% duty cycle, this corresponds to a maximum digitization rate of 12.8 .mu.sec. For conventional multi-slice imaging, 512 complex I/Q pairs can be typically acquired in 8 msec or 16 .mu.sec per point. Thus, it can be seen that a single filter chip may be sufficient for each coil channel 14', with some spare for overhead such as reloading the filter coefficients.

DEPR:
For three-dimensional imaging, the incoming data rates are the same as for multi-slice data, but another dimension of filtering is required. Since the data is acquired over many minutes with nearly 100% duty cycles, huge amounts of data are processed. It is therefore not practical to place temporary memory in front of each filter, and thus the data must be combined as fast as it enters. A three-dimensional image with 512 readout points and an 8 msec readout time would require 16 filter chips to keep up with the incoming data rate. A reduction in the filter kernel from 16 to 8 in the two phase encoding directions changes the data rate through the filter by a factor of four and this only four chips would be required. This may cause some degradation in the SNR (initial indications are not much) but the SNR will still be better than one could obtain without using the phased array.

DEPR:
A four-coil array was used to demonstrate the methods for single slice saggital imaging of the human spine. The array was made of 12 cm coils overlapping in a row in a manner similar to that shown in FIG. 4 of the '162 patent. The four coils 12 were placed beneath the patient in a linear array running in the vertical direction. Each coil 12 had its own receiver and

digitizer. The image FOV was 40 cm with a composite matrix size of 512.times.512 pixels. After the data was separately collected, it was combined in the time domain with filter kernel sizes ranging from 1.times.1 (a simple sum) to 9.times.9. For comparison, the data was also combined in the image domain.

DEPR:
Following the general procedure described in the '162 patent, the data from each coil was combined in the image domain by separately Fourier transforming the data from each coil 12, weighting each image by its corresponding field map image, and then summing the results. The image was then normalized into a uniform sensitivity image. FIG. 10A shows the resultant 512.times.512 saggital image of the spine.

DEPR:
To combine the data in the time domain utilizing the method of the present invention, the filter coefficients were determined by calculating the complex field map (magnitude and phase) for each coil over the full 512.times.512 image matrix. Expressions for the field maps were obtained by integrating the Biot-Savart Law over a conductor placed at the centerline of the coil. The complex field map for each coil was then inverse Fourier transformed into k-space and truncated to the desired size. To avoid ringing in the image, the resultant filter coefficients were windowed in two dimensions with a Hamming window of the type described in R. W. Hamming, Digital Filters, supra.

DEPR:
To avoid interference patterns in the image, the raw data was corrected for the different phases and gains of the NMR receiving channels. To do this, a transmit loop approximately 2 cm in diameter was placed about 4 cm above each surface coil. The transmit loop was driven from the local oscillator used by the NMR receivers and this was phase-locked to the receiver. The resultant signal amplitude and phase measured at the output of the receivers were used to calibrate each channel.

DEPR:
Using equation (6), the truncated filter kernel was then convolved with the phase corrected NMR data. The results were then summed over the coils and Fourier transformed. To obtain a uniform sensitivity image, the filter kernel was convolved with the conjugate of the filter kernel and summed over the coils. This normalization map was also Fourier transformed and divided pixel-by-pixel into the image.

DEPR:
FIGS. 10B, 10C and 10D show the results of a simple sum, a 5.times.5 point and 9.times.9 point kernel. The simple sum image (FIG. 10A) corresponds to an image obtained using a single large coil. As expected, the simple sum image had poor SNR (approximately two times lower than the point-to-point weighted sum image of FIG. 10A) and poor suppression of motion and wraparound artifacts. The 5.times.5 kernel image showed significant improvement. The 5.times.5 image (FIG. 10C) was developed by passing the time dependent data from each coil 12 through a two-dimensional filter with a 5.times.5 kernel before summing. A faint wraparound artifact is visible at the top. The 9.times.9 image (FIG. 10D) was constructed similarly using a 9.times.9 kernel. The 9.times.9 result had almost the same quality as the composite image (FIG. 10A) developed using the image domain techniques, except for minor differences near the edges of the image.

DEPR:
The differences of the edges between the images obtained using time domain (FIGS. 10B and 10D) and image domain (FIG. 10A) methods are due to the wraparound of the filter. Ideally, the filter corresponding to the coil at the bottom of the image should have no significant contribution at the top of the image. However, a 9.times.9 filter can be made to roll off in only about 1/9th of the image FOV and, thus, some wraparound is unavoidable. Combining the data in the image domain, however, allows one to filter to the nearest pixel or 1/512 of the image FOV. To obtain exactly the same result in the time domain would required a 512.times.512 filter kernel.

DEPR:

In the above example, the filter coefficients were determined for the time domain filter hardware by calculating the RF magnetic field for each pixel in each slice for each coil. The results were then Fourier transformed, truncated, and then windowed. This method may not be fast enough, however, to be practical in the clinical environment. The filter coefficients are a function of the slice position and the operator of the NMR imaging system selects the slice orientation and the number of slices on a patent-by-patient basis. Within a minute or two after the selection, the NMR instrument needs to be ready to take data. Using the method of the example, to determine the coefficients for 30 slices of 512.times.512 pixel images using the four-coil array, 120 two-dimensional 512.times.512 complex inverse Fourier transforms would be required. If each transform takes 3 seconds on an array processor, this part of the computation would take 6 minutes. The calculation of the magnetic fields would probably add a few more minutes, thus creating a built-in delay of about 10 minutes, which might be unacceptable. A more rapid means of calculating the filter coefficients from the known positions or the RF receiving coils may be needed.

DEPR:
There are a couple of approaches to speeding up the process of filter calculation. One approach is to precalculate and store filter coefficients for common slice locations, but this might be too restrictive. Another approach involves a Fourier transform on a smaller grid. Since the resultant set of filter coefficients in k-space will be truncated, it is not necessary to sample the RF magnetic field at each and every pixel in the image before inverse Fourier transformation. The matrix size for coefficient calculations can thus be reduced from, say, 512 to perhaps 50 or fewer pixels. In the above 30 slice calculation, this would decrease the time by a factor of 100.

DEPR:
A further approach involves the determination of the full three-dimensional representation of a particular coil's RF magnetic field for a known position of the coil. This is done for each vector component, i.e., B.sub.x, B.sub.y, and B.sub.z, of the magnetic field. Each component of the RF magnetic field is then inverse Fourier transformed, truncated, and saved on disk for later use. Since rotations in real space are simple rotations in k-space and translations in real space are phase shifts in k-space, the set of filter coefficients can be derived for any coil location or orientation from this original stored set. For three-dimensional imaging, the field maps can be simply rotated and translated in k-space and then windowed to the desired size. For multi-slice data, the three-dimensional k-space data can be first rotated and then translated according to the slice position. The data can then be collapsed into two dimensions before windowing. Such methods involve relatively simple operations on small matrices, so they can be accomplished quickly.

DEPR:
In yet another approach, the NMMR image data itself can be used as the basis for determination of the filter coefficients. This is analogous to a sum-of-squares image approach and relies on the fact that the image itself is a measure of the coil sensitivity. By acquiring the center of k-space first, the raw data itself (actually its conjugate) becomes the filter coefficient for each channel.

DEPR:
There has thus been described a method for combining the data from the separate coil channels of an NMR phased array in the time domain using filters to produce a composite image having high SNR throughout the image. When compared with methods that combine the data in the image domain, substantial reductions in the reconstruction time and the amount of memory required in the NMR imaging system are realized. In this way, systems using many coils can be made more practical.

DEPL:
where the * denotes the complex conjugate and C is an overall scale factor. The complex conjugate enters equation (2) because increasing angles of the RF magnetic field are defined to be positive in the direction of rotation of the nuclei. Greater angles of the RF magnetic field correspond to time delays (negative phase shifts) and thus the NMR signal is proportional to the complex conjugate of the RF magnetic field.

DEPC:
Combination in the Image Domain

DEPC:
Combination in the Time Domain

CLPR:
1. A method for combining NMR response data of a sample from a plurality of RF receiver coils of an NMR phased array in the time domain to form a composite NMR image, comprising the steps of:

CLPR:
8. The method of claim 1 wherein the step of convolving includes the steps of obtaining an initial NMR image representation of the sample from each of the RF receiver coils and substituting the initial NMR image representation for the time domain representation of each receiver coil.

CLPR:
9. The method of claim 1 wherein the NMR composite image comprises a spectroscopic image of the sample.

CLPR:
10. A method for combining NMR response data from receiver coils of an NMR phased array in the time domain using filtering to form a composite NMR image having good overall SNR resolution, comprising the steps of:

CLPR:
11. Apparatus for receiving and combining NMR response data from a plurality of RF receiver coils of an NMR phased array to form a composite NMR image of a sample, comprising:

CLPV:
(a) receiving at each of the RF receiver coils a different one of a plurality of NMR response signals, each of the signals being evoked from a portion of the sample within a field of view of a respective one of the receiver coils;

CLPV:
(d) combining the signals obtained by the step of convolving on a time domain point-by-point basis to produce a time domain representation of the composite NMR image of the sample.

CLPV:
(b) substantially simultaneously receiving at each one of the coils a different one of a plurality of NMR response signals, each evoked from a portion of the sample within the field-of-view of that coil;

CLPV:
(f) combining the multiplied temporarily stored signals of each coil, on a time domain point-by-point basis, to produce a composite NMR image of the sample.

CLPV:
a plurality of receiver circuits each connected to a respective one of the coils for receiving a corresponding NMR response signal, each response signal being evoked from a portion of the sample within the field-of-view of the response one of the coils;

CLPV:
means connected to said temporary storing means for combining said convolved series to provide a time domain representation of a composite NMR image of the sample.

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Search Results - Record(s) 1 through 5 of 5 returned.

☐ 1. Document ID: US 20010003423 A1

L30: Entry 1 of 5 File: PGPB Jun 14, 2001

PGPUB-DOCUMENT-NUMBER: 20010003423
PGPUB-FILING-TYPE: new-utility
DOCUMENT-IDENTIFIER: US 20010003423 A1

TITLE: Magnetic resonance imaging

PUBLICATION-DATE: June 14, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Wald, Lawrence L.	Cambridge	MA	US	

US-CL-CURRENT: 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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☐ 2. Document ID: US 6181134 B1

L30: Entry 2 of 5 File: USPT Jan 30, 2001

US-PAT-NO: 6181134
DOCUMENT-IDENTIFIER: US 6181134 B1

TITLE: Magnetic resonance imaging of the distribution of a marker compound without obtaining spectral information

DATE-ISSUED: January 30, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wald; Lawrence L.	Cambridge	MA		

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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☐ 3. Document ID: US 6144873 A

L30: Entry 3 of 5 File: USPT Nov 7, 2000

US-PAT-NO: 6144873
DOCUMENT-IDENTIFIER: US 6144873 A

TITLE: Method of efficient data encoding in dynamic magnetic resonance imaging

DATE-ISSUED: November 7, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Madore; Bruno	Redwood City	CA		
Glover; Gary H.	Stanford	CA		
Pelc; Norbert J.	Los Altos	CA		

US-CL-CURRENT: 600/410; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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4. Document ID: US 6080164 A

L30: Entry 4 of 5   File: USPT   Jun 27, 2000

US-PAT-NO: 6080164
DOCUMENT-IDENTIFIER: US 6080164 A

TITLE: Versatile stereotactic device

DATE-ISSUED: June 27, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Oshio; Koichi	Tokyo			JPX
Panych; Lawrence P.	Brookline	MA		
Guttmann; Charles R. G.	Brookline	MA		

US-CL-CURRENT: 606/130

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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5. Document ID: US 5245282 A

L30: Entry 5 of 5   File: USPT   Sep 14, 1993

US-PAT-NO: 5245282
DOCUMENT-IDENTIFIER: US 5245282 A

TITLE: Three-dimensional magnetic resonance imaging

DATE-ISSUED: September 14, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Mugler, III; John P.	Charlottesville	VA		
Brookeman; James R.	Charlottesville	VA		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference

KWIC | Draw Desc | Image

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Term	Documents
PLURALITY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2010088
PLURALITIES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	14034
PLURALITYS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
MULTIPLE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	900481
MULTIPLES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	29484
MULTI.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	625373
MULTIS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	40
SLICE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	47536
SLOUSE	0
SLOUSES	0
(29 AND (SLICE WITH (MULTI OR MULTIPLE OR PLURALITY))).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	5

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Search Results - Record(s) 1 through 1 of 1 returned.

☐ 1. Document ID: US 5568400 A

L44: Entry 1 of 1 File: USPT Oct 22, 1996

US-PAT-NO: 5568400
DOCUMENT-IDENTIFIER: US 5568400 A

TITLE: Multiplicative signal correction method and apparatus

DATE-ISSUED: October 22, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Stark; Edward W.	New York	NY	10023	
Martens; Harald	Aas			NOX

US-CL-CURRENT: 702/85; 702/27

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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Term	Documents
MAGNETIC.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1038365
MAGNETICS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9279
RESONANCE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	189357
RESONANCES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	10347
MRI.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	11712
MRIS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	85
NMR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	92175
NMRS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	109
(39 AND (MRI OR (MAGNETIC ADJ RESONANCE) OR NMR)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

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L44: Entry 1 of 1 File: USPT Oct 22, 1996

DOCUMENT-IDENTIFIER: US 5568400 A
TITLE: Multiplicative signal correction method and apparatus

BSPR:
In image analysis the fundamental variable is usually distance in one or two dimensions although the two-dimensional Fourier transform, also known as the Weiner transform, and the Weiner spectrum which express the information in a two-dimensional spatial frequency domain are also prevalent. Multivariate images, such as three color video signals and many satellite images where each picture element is characterized by a multichannel "spectrum" and also images constituting a time sequence of information, provide additional dimensionality in the data. Alternatively, in image analysis, the images can be summarized into histograms showing distributions of various picture elements, where the variable is then a vector of categories, each representing a class of picture elements, e.g. various gray levels of pixels, or contextual classes based on local image geometry. For multivariate images, the additional multichannel information may be included in the contextual classification. Time information can likewise be included in the definition of the categories in the variable. The above descriptions of two-way images also apply to three-way tomographic images, e.g. in MRI and X-ray tomography.

BSPR:
A prior approach to minimizing these errors has been to omit those portions of the spectrum having large variability from the data used in the regression. This approach is sometimes difficult to apply, because it may require many trials and operator judgments, and it is only partially successful at best. In a variation of this approach, the range of the spectral data included in the average spectrum used to determine the offset and slope coefficients is restricted to the vicinity of a strong isolated spectral feature, such as a solvent absorption band, thereby limiting the magnitude of the residuals and improving the accuracy of the correction. This variation has been applied to correction of the effects of scattering within the specimen in transmission spectroscopy. In many cases, however, there is no strong isolated band available for determination of the multiplicative correction. A related problem occurs in measuring one material through another with the pathlength through each material unknown and variable. In either case, better means are needed to accurately separate additive and multiplicative effects.

BSPR:
The present invention includes using any linear multivariate estimator to determine the correction coefficients such as multiple linear regression, generalized least squares, maximum likelihood regression, robust regression, estimated best linear predictor, partial least squares, principal component regression, Fourier regression, covariance adjustment, or non-Euclidian distance measures.

BSPR:
The present invention also comprises, as option B, generating modified reference spectra of the interfering components that contain only those portions of the original reference spectra that are orthogonal to, and therefore uncorrelated with, one or more reference analyte spectra. The coefficients generated for these orthogonal spectra are not influenced by the presence or magnitude of analyte information contained in the raw data even if the analyte spectrum is not included in the modeling or the coefficient estimator does not inherently orthogonalize the components, so they may be a

more correct representation of the magnitude of the spectral effects of the interfering components. These more accurate coefficients are then used to scale the original reference spectra prior to subtraction from the input data and the correction proceeds as in option A. This option reduces or eliminates the error in analyte spectral contribution that would otherwise be caused by subtracting an incorrect amount of a spectrum which contains some information equivalent to analyte information.

DEPR:
Like the conventional Martens, Jensen, and Geladi multiplicative scatter correction (MSC), it seeks to normalize every input spectra to some reference "ideal" or "average" state by additive and multiplicative normalization, after having estimated the offset and slope parameters by some type of regression against some reference spectrum over some selected wavelength range, and this reference spectrum may be of the same kinds as those employed in MSC. However, it extends conventional MSC by explicitly modeling the effects of anticipated additive interferences and by optionally utilizing nonlinear modeling in deriving the additive and multiplicative normalization. This in turn improves the accuracy of the multiplicative correction, it allows removal of undesired interferants already at the multiplicative preprocessing stage, and it simplifies a causal understanding of the multiplicative correction and facilitates its interactive graphical optimization. It may also create interference reference spectra orthogonal to the analyte(s) spectra for use in the modeling to avoid the effects of intercorrelation between the interferant spectra and the analyte spectra which would otherwise cause inaccuracies in estimating the interferant coefficients and in the subsequent subtraction of their contribution to the spectral data being normalized.

DEPR:
The R.sub.ka and R.sub.kj reference spectra are often statistical estimates extracted from the measured data from sets of specimens, although directly measured spectra are also useful in many cases. Honig's spectral reconstruction (D. E. Honigs, G. M. Hieftje, and T. Hirschfeld, A New Method for Obtaining Individual Component Spectra from Those of Complex Mixtures, Applied Spectroscopy, 38(3), pp. 317-322, a copy of which being annexed hereto) provides a method for extracting spectra from a set of mixture specimens based on knowledge of the concentration values. Principal component analysis (PCA) and partial least squares (PLS) provide orthogonal sets of spectra representative of the variation in the data. Stark's method (U.S. patent application Ser. No. 07/319,450) provides reference spectra for previously unknown variations based On analysis of replicate data. In the simplest operation of the present invention, i.e. correction for offset and multiplicative factors, the primary requirement of R.sub.ka and R.sub.kj is that they reasonably span the variation of X.sub.ki so as to stabilize the modeling and spectral accuracy and specificity are not essential. For the more complex options, in which R.sub.ka and R.sub.kj are incorporated into the output data as corrections, the quality of R.sub.ka and R.sub.kj become more important. The accuracy and specificity of the R.sub.ka spectral data is particularly important in orthogonalization of R.sub.kj either explicitly or implicitly and when used for added weight as described below. The spectral information in R.sub.ka and R.sub.kj may be represented in various ways with respect to redundancy and collinearity, for example one individual vector for each phenomenon, several replicates or specimens, statistical summaries (averages, bilinear components, square root of covariance matrices, etc.), or rotated representations where some or all collinearities have been eliminated. In a preferred embodiment, redundancy is eliminated by averaging so that the number of vectors equals the number of phenomena being modeled.

DEPR:
In a preferred addition to the basic preferred embodiment described above and shown in FIG. 3, additional correction spectra [R.sub.ka *b.sub.ai] and [R.sub.ki *b.sub.ji] are formed by matrix multiplier 340 from the reference spectra and their associated coefficients that were also generated in coefficient estimator 320. A matrix multiplier to form the spectrum [R.sub.kn *b.sub.ni] for a single input spectrum i, consists of short term storage for the both inputs, a multiplication and summation circuit, an address sequencer which accesses the corresponding elements n of R.sub.kn and B.sub.nk and a second address sequencer which accesses the rows k of R.sub.kn and addresses the short term storage which keeps the resulting k.times.1 matrix. Matrix

multiplication is also a standard function of available array processors. This additional combined correction spectrum may be used directly by subtractor 333 to further correct Y.sub.ki, which becomes

DEPR:
The orthogonal component generator 360 provides for transformation of the reference spectra [R.sub.ks,R.sub.ka,R.sub.kj] into a new set of spectra, [P.sub.ks,P.sub.ka,P.sub.kj], some or all of which are orthogonal to each other. If the reference spectra are latent variables derived from a single PCA or PLS analysis, they are orthogonal by definition. If they are measured spectra of components or otherwise separately derived, they will generally be intercorrelated, which if severe enough may cause errors in the coefficient values or failure of the coefficient estimator to complete its operation. If orthogonal reference spectra are created, new reference spectra may be added without requiring complete recalculation by the coefficient estimator. Orthogonal reference spectra also minimize the number of operations required by the coefficient estimator to determine the coefficients.

DEPR:
In a preferred embodiment, the orthogonal component generator performs a Gram-Schmidt orthogonalization in accordance with ##EQU4## where I=the identity matrix

DEPR:
The variations of R.sub.ks are preserved in P.sub.ks, therefore the coefficient bs is not affected by the orthogonalization. Each succeeding R.sub.kn is then orthogonalized against the matrix formed by the preceding orthogonal P.sub.kn spectra, until all reference spectra are orthogonalized into matrix P.sub.kn =[1,P.sub.ks,P.sub.ka,P.sub.kj]. Each spectrum P.sub.kn comprises the residuals of the regression of R.sub.kn on the preceding orthogonal P.sub.kn. If a spectrum P.sub.kn is 0 or has only small values, it provides warning of dependence between spectra that could cause problems in coefficient estimation. In such case, the information is provided to the operator, or separate decision circuitry, to determine whether to delete the spectrum from the model, to downweight its importance, or to accept it without change. Orthogonalization processing may be performed solely for the purpose of generating this warning information. When full orthogonalization is chosen, the reference spectra input to matrix multiplier 340 are the P.sub.kj and P.sub.ka.

DEPR:
The orthogonal component generator and storage 360 comprises storage for P.sub.kn, the portion that is filled as the process proceeds comprising Z, storage for [Z'Z]-1, storage for the intermediate product Z(Z'Z)-1, storage for Z.sub.i, storage for the intermediate product Z'Z.sub.i, point by point multiply and sum logic, scalar inversion (1/a) logic, a subtractor, and the sequencer to select data from storage for processing, to control the processing sequence, and to direct storage of results. Circuit devices to perform these functions include the Intel 80287 math coprocessor for hardware implementation of the arithmetic functions, CMOS static ram chips (e.g. 4 parallel Motorola MCM6226-30 128K.times.8) to provide 32 bit resolution in storage of the digital data, and standard programmable array logic devices (PAL's) combined with a clock and counter as the sequencer. Each matrix element is acted on in sequence in accordance with the hardware logic. The required functions can also be obtained with a standard array processor operated in sequential fashion by the sequencer.

DEPR:
The contents of Z[Z'Z].sub.-1 is the transpose U' of U=[Z'Z]-1Z' which is useful in finding multiple linear regression coefficients by matrix multiplication.

DEPR:
If even this degree of reference spectrum modification is undesireable, the orthogonal component generation is bypassed and the original reference spectra are passed to the coefficient estimator.

DEPR:
In the linear case, taking R.sub.kn =[1,R.sub.ks,R.sub.ka,R.sub.kj], a matrix

where each row represents observations at a value k of the spectral variable and each column is a reference spectrum R.sub.kn incorporated in the model, coefficient estimator 320 fits X.sub.ki to Rnk by some method, minimizing the residuals e.sub.ki in

DEPR:
Methods for achieving this linear modeling include generalized least squares, maximum likelihood regression, robust regression, estimated best linear predictor, partial least squares, principal component regression, Fourier regression, covariance adjustment, and others. For example, generalized least squares with generalized inverse models X.sub.ki by

DEPR:
A second preferred embodiment of the coefficient estimator which avoids matrix inversion is a principal components regression (PCR) device, which requires no pretreatment of R.sub.kn and no matrix inversion.

DEPR:
In the case of nonlinear modeling, the coefficient estimator 320 becomes more complex as each nonlinear coefficient becomes a vector of length k. C.sub.ki and D.sub.ki are therefore matrices containing a number of coefficient vectors that depends on the form of the nonlinear model.

DEPR:
A preferred embodiment uses the coefficient estimator 320 illustrated in FIG. 4 which employs Taylor series linearization. The model response generator 321 calculates the vector F from the reference spectra Rkn, the present value Ari of the coefficients being generated by the iterative process, the variable k and the input spectral data Xki. This operation involves matrix multiplication and summation in accordance with the appropriate form of model as discussed above. The set of coefficients Ari, comprising cni of Cki, dni of Dki, and bni, are initially stored in coefficient Aq-1 storage 322a. They may be modified by means of adder 322b through addition of a weighted correction wGq or of increment dAr to one of the coefficients at a time to create the present values stored in coefficient Aq storage 322c and used by model response generator 321. The remaining functions will become obvious from the following description of the operation.

DEPL:
X is the spectral ordinate, e.g. absorbance, fluorescent energy, or pixel intensity or relative count, representing the measurement system response. The subscript i denotes the object or specimen while subscript k (k=1,2, . . . ,K) is the spectral variable. k may be representative of a single dimension, e.g. wavelength in optical spectroscopy, two dimensions, e.g. time and wavelength in GC-IR measurements, or more depending on the measurement technology utilized. As used here, names of matrices are capitalized (e.g. R.sub.kj) and a matrix may comprise a single row Dr a single column of elements, for example X.sub.ki, Y.sub.ki, and R.sub.ks are individual spectra represented by single column matrices (vectors) of length K. Sets of multiple spectra form matrices (e.g. R.sub.ka, R.sub.kj, and Q.sub.km). Quantities that only exist as vectors or scalers are not capitalized.

DEPL:
Therefore, a general form for C.sub.ki and D.sub.ki can be described as a product of these series, ie a new series in terms of the powers of k and X and their cross products, removing redundant constants and terms in k or X and normalizing so that the linear magnitude information is kept in [R.sub.kn *t.sub.ni] and C.sub.ki and D.sub.ki carry only the information relating to the nonlinearity. C.sub.ki and D.sub.ki are matrices containing k rows, and as many columns as required for the number of terms in the appropriate power series approximations.

DEPL:
Z comprises orthogonal columns therefore [Z'Z] is diagonal of size (i).times.(i) and determining [Z'Z].sup.-1 is trivial by inversion of the individual elements.

DEPL:
With this procedure, R.sub.ka may be omitted in the estimation of the b.sub.kj

coefficients without causing errors in their determination. This method has the advantage of only removing analyte related information from the reference spectra, thus the analyte spectrum is unaffected, the interferant spectral shapes are minimally affected, and the coefficients have physical interpretations. In this case, the correct input to the matrix multiplier 340 is R.sub.kj rather than P.sub.kj, to properly subtract the portion of R.sub.kj correlated with R.sub.ka. Implementation of this digital logic requires only a subset of the functions described previously.

DEPL:
where [ ].about. means a generalized inverse and where covariance matrix v(i) can be iteratively updated based on the previous fit for this specimen i.

DEPL:
can be precomputed externally and stored with the reference spectra, thereby minimizing the requirements on the data normalizer 300. If full rank Gram-Schmidt orthogonalization is used, U.sub.nk is available from that process. In either case the calculation of the coefficients of the linear model involves a simple matrix multiplication. A matrix multiplier for U.sub.nk and X.sub.ki consists of short term storage for one or both inputs, a multiplication and summation circuit, an address sequencer which accesses the corresponding elements k of U.sub.nk and X.sub.ki and a second address sequencer which accesses the rows n of U.sub.nk and addresses the short term storage which keeps the resulting b.sub.ni values. In the more general case of multiple linear regression, matrix [R'R] must be formed and inverted to obtain [R'R].sup.-1 prior to matrix multiplication by R' to obtain U.sub.nk. This function can readily be accomplished with an available array processor and suitable logic sequencer.

DEPL:
where X=(x.sub.ik) is the matrix of spectral ordinates for i=1,2, . . . N objects, k=1,2, . . . ,K wavelengths, T=(t.sub.il) is the matrix of scores for objects i, bilinear factors l1=1,2, . . . ,L obtained from some bilinear model (e.g. principal component analysis, partial least squares, etc.), P=(p.sub.kl) are the loadings for objects i on bilinear factors l, and E=(e.sub.ik) are the residuals between data X and model T*P'. The loadings P can then be decomposed into a function of a reference spectra r=(r.sub.k) (e.g. the Bean of the X data) and a matrix G=(g.sub.km) spanning the spectra for analyte and interference phenomena m=1,2, . . . ,M:

DEPL:
where d=(d.sub.1) and h=(h.sub.1) are vectors of length L, 1 is a vector of ones of length K, C=(c.sub.lm) is a matrix of regression coefficients of size L.times.M which quantifies the analyte and interference contributions, and F=(f.sub.lk) contains the residual loadings with the multiplicative, analyte, and interference phenomena removed. d, h, and C can be estimated by regression of P' on r, 1, and G by some method (e.g. weighted least squares). C*G' could be reduced in size by elimination of effects if the relative size of the chemical or interferent effects are small

DEPL:
where U=is a matrix of eigenvalues and V' is a matrix of eigenvectors. By substitution,

DEPL:
The product T*U produces offset and interferent corrected scores and V' is the matrix of corresponding spectra loadings associated with the corrected scores.

DEPL:
where S.sup.-1 is the inverse of S. The fully corrected score matrix W is found in a similar fashion,

DEPL:
W=(W.sub.il) is the matrix of the offset, multiplicative, and interferent corrected scores which can be used as regressors in additive mixture models etc.

DEPV:

Z=the matrix of vectors already transformed,

DEPV:
Z.sub.i T=transformed vector orthogonal to vectors already in Z.

DEPV:
1. Let X=(x.sub.ik) be the matrix of spectral ordinates for i=1,2, . . . N objects, k=1,2, . . . ,K wavelengths. The multiplicative effect is modeled from the spectral data using a standard multiplicative scatter correction (i.e., avoiding the use of components for practicing the present invention) yielding the corrected spectral data Z.

DEPV:
5. Construct a new matrix of corrected spectra Z from X and repeat step 2, step 3, and step 4 until convergence Occurs.

DEPV:
Let S be the diagonal matrix containing the elements of the product T*d. The fully corrected spectral data are found by

CLPR:
11. The method of claim 6 wherein said model includes a covariance adjustment technique.

CLPR:
18. The method as in claim 17 including the steps of generating modified reference spectra of interfering components that contain only those portions of original reference spectra of the interfering components that are orthogonal to, and therefore uncorrelated with, at least one reference analyte spectrum, using the coefficients generated from said orthogonal reference spectrum to scale the original reference spectra prior to subtracting the scaled spectra from the data.

CLPL:
where X=(x.sub.ik) is the matrix of spectral ordinates for i=1,2 . . . N objects, k=1,2 . . . , K wavelengths, T=(t.sub.il) is the matrix of scores for objects i, l=1,2 . . . L representing bilinear factors Obtained from a bilinear model, P=(P.sub.ki) are the loadings for objects i on bilinear factors l, and E=(e.sub.ik) are the residuals between data X and model T*P';

CLPL:
where X=(x.sub.ik) is the matrix of spectral ordinates for i=1,2 . . . N objects, k=1,2 . . . , K wavelengths, T=(t.sub.il) is the matrix of scores for objects i, l=1,2 . . . L representing bilinear factors obtained from a bilinear model, P=(P.sub.ki) are the loadings for objects i on bilinear factors l, and E=(e.sub.ik) are the residuals between data X and model T*P';

CLPL:
where X=(x.sub.ik) is the matrix of spectral ordinates for i=1,z . . . N objects, k=1,2 . . . , k wavelengths, T=(t.sub.il) is the matrix of scores for objects i, l=1,2 . . . L representing bilinear factors obtained from a bilinear model, P=(P.sub.ki) are the loadings for objects i on bilinear factors l and E=(e.sub.ik) are the residuals between data X and model T*P';

CLPL:
where X=(x.sub.ik) is the matrix of spectral ordinates for i=1,z . . . N objects, k=1,2 . . . , k wavelengths, T=(t.sub.il) is the matrix of scores for objects i, l=1,2 . . . L representing bilinear factors obtained from a bilinear model, P=(P.sub.ki) are the loadings for objects i on bilinear factors l and E=(e.sub.ik) are the residuals between data X and model T*P';

CLPV:
5) constructing a new matrix of corrected spectra Z from X; and

CLPV:
decomposing the loadings into a function of a reference spectral and, optionally, a matrix of spectral components for analyte and interference phenomena;

CLPV:
decomposing the loadings into a function of a reference spectrum and, optionally, a matrix of spectral components for analyte and interference phenomena;

CLPV:
determining an S.sup.-1 factor related to a matrix containing T*d for multiplicative effects from the new spectral scores T and using a factor U from calibration relating to additive and interferent effects to find

CLPV:
W=S.sup.-1 *T*U; where W is the matrix of the additive, multiplicative and interferent corrected scores which can be used as regressors in additive mixture models;

CLPV:
fifth means for providing a signal representing the construction of a new matrix of corrected spectra Z from X;

CLPV:
means for providing a signal representing decomposed loadings which have been decomposed into a function of a reference spectra and, optionally, a matrix of spectral components for analyte and interference phenomena;

CLPV:
means for providing a signal representing decomposed loadings which have been decomposed into a function of a reference spectra and, optionally, a matrix of spectral components for analyte and interference phenomena;

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☐ 1. Document ID: US 5572125 A

L49: Entry 1 of 1 File: USPT Nov 5, 1996

US-PAT-NO: 5572125
DOCUMENT-IDENTIFIER: US 5572125 A

TITLE: Correction and automated analysis of spectral and imaging data

DATE-ISSUED: November 5, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Dunkel; Reinhard	Salt Lake City	UT	84102	

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference   KWIC | Draw Desc | Image

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MATRIXES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	5559
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☐ 1. Document ID: US 5983251 A

L55: Entry 1 of 1 File: USPT Nov 9, 1999

US-PAT-NO: 5983251
DOCUMENT-IDENTIFIER: US 5983251 A

TITLE: Method and apparatus for data analysis

DATE-ISSUED: November 9, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Martens; Harald Aagaard	Oslo			NOX
Reberg; Jan Otto	Langhus			NOX

US-CL-CURRENT: 708/203

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

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EIGENVALUE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1210
EIGENVALUES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1244
(54 AND (EIGENVECTOR OR EIGENVALUE)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

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